Estimation of arterial pulse wave velocity
with a new improved Tissue Doppler method

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Abstract- **The mechanical properties of the large arteries are important determinants in the circulation physiology. Quantities associated with vessel wall elasticity are of particular interest since they offer a possibility to separate diseased arteries from healthy. Pulse wave velocity (PWV) is a local measure of the arterial wall elasticity. This paper describes a new improved method for non-invasive local PWV estimation. The PWV estimation is based on arterial wall movements, which is detected by Tissue Doppler Imaging (TDI), a color Doppler optimized for low velocities. The TDI facility allows the artery to be measured at several positions simultaneously, which improves the PWV estimation considerably. The method is validated with respect to operator repeatability and operator reproducibility in a limited clinical study.**

Keywords- **Pulse wave velocity, PWV, non-invasive, Tissue Doppler Imaging**

INTRODUCTION

The mechanical properties of the large arteries are important determinants in the circulation physiology [12] Quantities associated with vessel wall elasticity are of particular interest since they offer a possibility to separate diseased arteries from healthy [11].

If the elastic properties of human arteries are generalized, some broad outline becomes evident. One of the most striking facts is that arteries become stiffer with age. In the abdominal aorta, arterial stiffness increases in a linear fashion by age among women whereas men show a faster exponential increase [16]. Furthermore, peripheral arteries are stiffer than more central located [14,9]. The age related stiffening is more pronounced in central arteries than in peripherals [1]. The stiffening is also more evident among individuals with hypertension and among individuals on high sodium diet [1,2].

The pulse wave velocity (PWV) is related to the wall elasticity according to the Moens-Korteweg formula (1), there E is the wall elastic modulus, h the wall thickness, ρ the blood density and R the lumen radius. Unfortunately the formula is of limited use for elasticity estimation due to the uncertain wall thickness determination. However, the pulse wave velocity itself may be used to quantify the arterial wall. This concept gives an estimate of the effective wall elasticity experienced within the lumen rather than the elastic modulus of the arterial wall itself.

$$c_0 = \sqrt{Eh/2R\rho} \qquad (1)$$

The basic idea in PWV estimation is to record a hemodynamic quantity at two locations simultaneously. The delay between the two waveforms in combination with the measuring distance permits calculation of the wave velocity. Bramwell and Hill performed one of the first studies in this area on excised arterial segments in 1922 [5]. To facilitate measurements, arteries were filled with mercury to reduce the propagation velocity. The development of pressure catheters offered a possibility to record pressure waves accurately with minimal disturbances in the circulation system [10].

For a more general use of the PWV concept, non-invasive techniques are clearly desirable. If the status in peripheral arteries is the main interest and average values over larger areas are acceptable, techniques utilizing various types of palpation are suitable. The arterial pulse is detectable on extremities through pressure, strain gauge or mercury ring sensors. These techniques are however less suitable for more selective measurements.

Local PWV measurements in specific arteries are simplified if the system depicts the target area. This facility in combination with PWV estimation is offered by MRI and ultrasonic scanners. Since the MRI technology is capable of recording blood flow through arbitrary scan planes, the PWV may be estimated [13]. One of the largest technical limitations of MRI as PWV estimator is the low data capture rate. To achieve an adequate temporal resolution, consecutive sampling of several heartbeats with ECG trigging is required. Recent research has however reduced recording time considerably [4]. An ultrasound scanner equipped with Doppler and ECG trigger may be used in basically the same way. As with MRI, flow waveforms in the different recording sites are not captured simultaneously but compounded off-line with the ECG signal as temporal reference. More specialized ultrasonic systems for PWV estimations allow the two recording sites to be measured simultaneously [3, 6, 7, 12].

Non-invasive ultrasonic techniques for local PWV estimation measure either wall pulsation [3] or wall velocity [6]. The two quantities are interchangeable since a derivation of the pulsation gives the velocity and an integration of the velocity gives the wall pulsation.

The aim of this study was to evaluate a new improved algorithm of the method *PWV estimation with TDI* described by Eriksson et al. [7]. The evaluation was performed *in vivo* and the study concentrated on operator repeatability and operator reproducibility.

METHODOLOGY

The PWV was estimated from the arterial wall motion over a defined segment. The wall motion was detected by Tissue Doppler Imaging (TDI), a color Doppler optimized for low velocities. The TDI facility allows the artery to be measured at several positions within same scan, with a short delay between the positions of exploration. The TDI information was

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Abstract

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provided by an HDI®-5000 ultrasonic system. The system was equipped with an L12--5 linear array (6.0 MHz Doppler) and run in the *Peripheral Vascular/Arterial* application. The data was stored in cineloops as consecutive frames consisting of TDI color information overlapped on the grey scale image, Fig. 1. The cineloops were transferred to a PC through the Research Link® for post processing. The cineloops containing pre scan-converted data were visualized on the PC in HDI-Lab, an image processing software designed for off-line cineloop analysis.

The settings of the ultrasonic system were optimized for PWV estimation and based on the outcome of the *in vitro* study [7]. To avoid aliasing, the TDI velocity range had to be extended compared to the previous study. The wider velocity range gave a considerably higher frame rate that reduced the number of heartbeats covered by the acquisition memory. To increase the number of beats, frame rate was reduced by a higher sensitivity. The system settings are given in Table I.
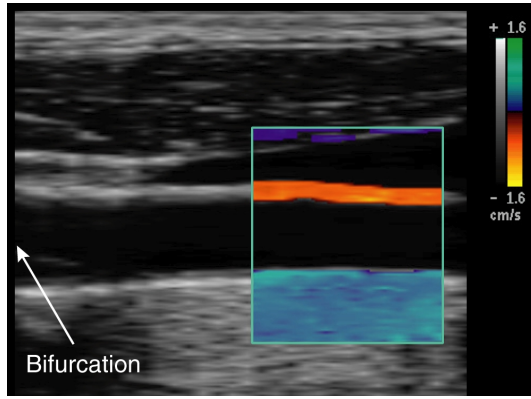

Fig. 1. Human carotid artery during expansion. TDI color box is placed to the right so the bifurcation onset remains in view as a reference.

TABLE I
SYSTEM SETTINGS

TDI vel. range [mm/s]	±16.0
Frame rate [Hz]	114
Frames	>790
TDI lines	14
TDI box width [mm]	16.8
TDI sensitivity [pulses]	5
Scan velocity [m/s]	17.2

Signal processing

The basic idea of the implemented algorithm was to detect the differential vessel wall velocity at multiple sites and estimate the wave delay along the artery. The vessel wall velocities were extracted from tissue motion given by the TDI information. The near and far vessel walls were defined by two manually adjusted region-of-interests (ROIs) Fig. 2A. Starting from the lumen, the first non-zero velocity sample within the ROI was regarded as a valid wall motion. To reduce noise, spatial averaging over ten samples (approximate 2 mm) was used in the axial-direction. Since the sample facing the lumen was considered as the best estimate, linear weighting was used in the averaging process. The first sample was given the weight of one and the last the weight of 1/10. Samples containing

dropouts and aliased data were excluded and corrected, respectively. The differential wall velocity was obtained by subtracting the far wall from the near wall. The wall velocity estimation was performed for each TDI line and gave a set of velocity traces as shown in Fig. 2A. The propagation delay along the vessel was estimated by crosscorrelating consecutive velocity traces, Fig. 2C. The delay was given by the peak shift in the correlation function. Since the intrinsic temporal resolution was more than a magnitude lower than typical shifts from adjacent TDI-lines, interpolation was required. A suitable interpolation factor was determined by comparing different factors with a reference (x1024). Based on the information in Fig. 3 an interpolation factor of 512 was chosen. To reduce computational time, a fraction of the interpolation was performed on the correlation function, Fig. 2D. A relation of 4:128 reduced the computational time by a factor of 50+ and gave virtually the same result as the one-step implementation, Figure 4. To improve the estimation, the correlation process was performed on all possible trace combinations, Fig. 2E.
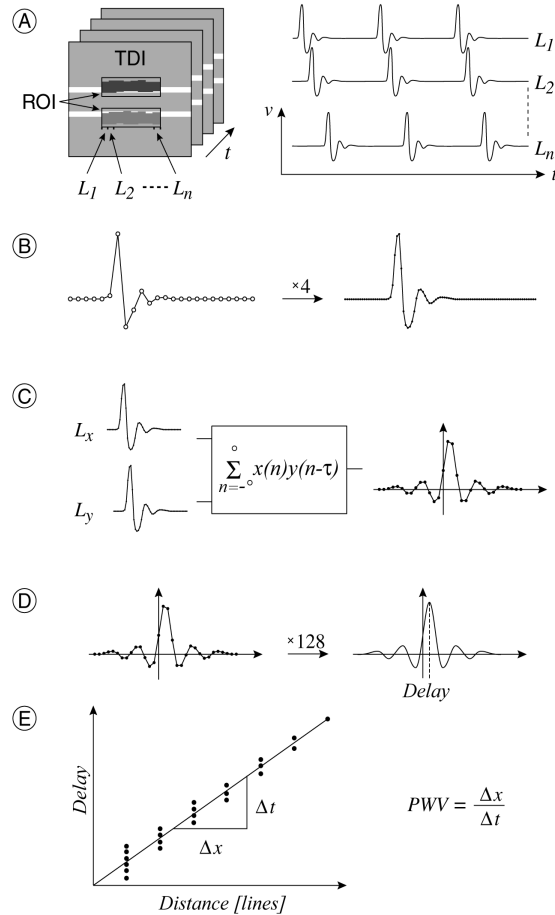

Fig. 2. Summarized description of the signal processing.
A) The vessel wall velocities were extracted from the tissue motion given by the TDI information. Two region-of-interests, ROIs, defined the desired wall segments. The velocity was interpolated (B) before crosscorrelation (C). The delay was determined from the interpolated correlation function (D). The pulse wave velocity (PWV) was estimated from the delays between the velocity traces (E).

Experimental group

The experimental group consisted of six non-smoking males with no known vascular defects. A physical description of the group is given in Table II. The PWV was measured in the right common carotid artery 20 mm below the bifurcation. The TDI box was placed in the left side of the image to maintain the bifurcation within the field of view, Fig. 1. The subjects were in spinal position and had at least 20 minutes rest before the examination. Each subject was examined by two experienced operators.

TABLE II
DESCRIPTION OF THE EXAMINED GROUP

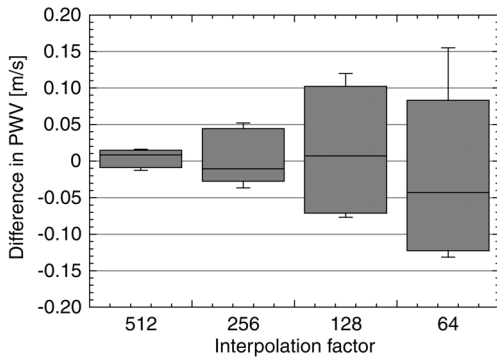| Subject	Age [years]	Height [cm]	Weight [kg]	Pressure [mmHg]
A	26	180	87	120/60
B	24	176	68	135/85
C	27	179	64	130/75
D	25	189	75	130/70
E	31	174	84	150/80
F	28	180	73	125/65

Fig. 3. Several recordings were processed with different interpolation factors. The results obtained with a factor of 1024 were defined as reference and subtracted from the other results. The factor 512 was chosen as the best compromise between resolution and required computational time.
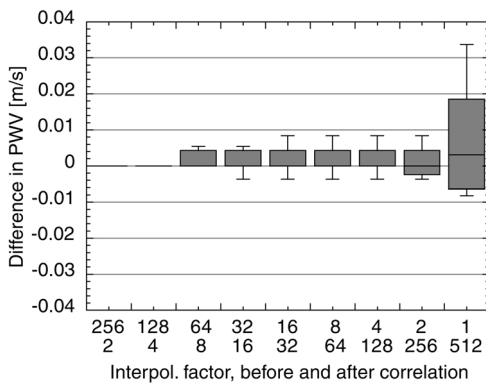

Fig. 4. To reduce computational time, interpolation was performed before and after the crosscorrelation. The 4:128 split reduced the computational time by a factor of more than 50 and gave virtually the same result as one-step implementation.

Experimental design and statistics

The evaluation was performed as a factorial experiment with operator and subject as factors. The experimental design was balanced and 7 replicates were obtained at each level. The study was not completely randomized since the replicates were recorded consecutively and all measurements on a subject were performed at one occasion. The operator order was however randomized.

The experimental outcome was evaluated through hypothesis testing. The H_0-hypotheses were associated with no operator effect, no subject effect and no interaction between operator and subject.

RESULTS

The estimated pulse wave velocities are shown in Fig. 5. The mean PWV ranges from 5.02 -- 11.24 m/s (subject C and F). The lowest standard deviation is found in subject E (0.65 m/s) and the highest in subject B (1.56 m/s). The overall mean of the examined group is 7.43 m/s.

The result of the statistical analysis is given in Table III. The operator has no significant effect on the PWV. There is furthermore no interaction between operator and subject. The subject has a significant effect on the PWV.

The repeatability of the method (expected variation when one operator performs multiple recordings on the same subject) was determined to 1.31 m^2/s^2. The reproducibility (additional variation caused by the operator) was 0.08 m^2/s^2.
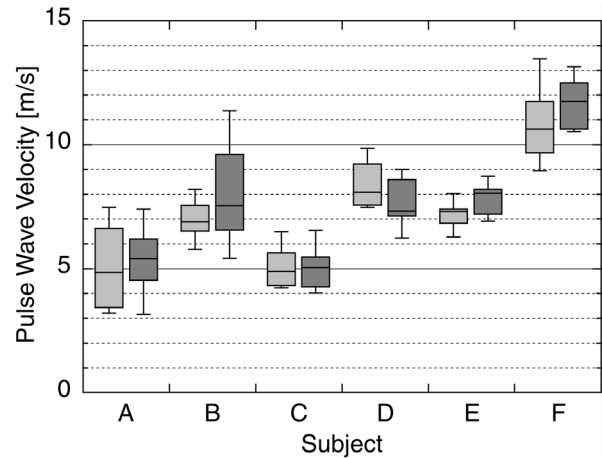

Fig. 5. Estimated pulse wave velocity in subject A to F. The different box fillings represent operator 1 and operator 2.

TABLE III
OUTCOME OF THE STATISTICAL ANALYSIS

Factor	Operator	Subject	Op · Subj
p-value	0.198	< 0.001	0.289

DISCUSSION

As shown by the evaluation, there is no significant operator effect. The relatively high repeatability indicates however a considerable recording variability. The obtained variability consists of two components; biological variations and method inadequacies. The amount of physiological PWV variations is difficult to estimate since no reference method was used. We believe that the variability in our study could not entirely be explained by physiological variations. This hypothesis is based on the fact that the human carotid artery has a limited capability for vasoactive work due to the reduced amount of smooth muscle cells [8]. On the other hand, an actual change in wall elasticity is enlarged in the PWV by the quadratic relationship (1).

The outcome of an *in vitro* study represents the method performance under ideal conditions. At comparable settings the *in vitro* evaluation [7] performed more than a magnitude better in reproducibility. By comparing the velocity traces in the two studies, *in vivo* data was found to have a considerably higher noise level. The vessel phantom used in the *in vitro* study (an elastic tube surrounded by water) produced well-defined echoes that gave a low noise level in the estimated wall velocities. In an *in vivo* situation echoes are less distinct which makes it more difficult to maintain an optimal vessel alignment. The lowest noise levels were found in subject C and E, which also showed the lowest variability. High quality recordings are consequently required to achieve low variability. The relatively high repeatability in this study would have been considerably lower if every recording had been inspected and accepted during the examination. This possibility requires some sort of quality factor to grade the recording objectively. Furthermore, this study shows the value of repeated measurements in TDI based PWV estimations.

This study was performed on the carotid artery but several other arteries are of interest. To obtain an accurate estimate of the pulse wave velocity a number of criteria must be fulfilled. The most obvious is that the vessel is detectable. The scan head used in this experiment had a working range of 5--12 MHz, which makes it suitable for superficial arteries. To examine deeper laying vessels such as the abdominal aorta a low frequency scan head is required. The larger depth might however reduce the maximum achievable frame rate, vital for accurate PWV estimation. Large peripheral arteries such as femoral, popliteal, brachial and radial offer, at present state, other difficulties. These arteries are generally stiffer than carotid and have a lower wall pulsation thereby. The available velocity ranges in TDI in combination with the limited dynamic range result in a strong velocity quantization. A lower and for this application more suitable velocity range would unfortunately also reduce the frame rate. Furthermore, the lower velocity range reduces the scan velocity to approximately 7 m/s, which is about half the value found in human muscular arteries [15]. PWV estimation is still possible but the variance tends to increase.

CONCLUSION

The described method offers a possibility to non-invasively estimate PWV in human arteries. In contrast to traditional non-invasive techniques that give an average value over a large distance, this method measures the PWV locally. The measurements are reproducible and the operator has no significant effect. In present state the system is restricted to superficial highly pulsatile arteries but further technical development will extend its capability.

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